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Multichannel Cuff Electrodes for Peripheral Nerve Stimulation and Recording

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***Abstract*—In the development of neuroprostheses to restore sensory and motor function to disabled patients the choice of the electrodes to be used remains an important consideration. The optimal electrode design should be minimally invasive and be capable of recording or stimulating selectively a large number of nerve fibers. Additionally, the electrodes should be capable of delivering stimulation within electrochemically safe limits. Here we report on the use of a multi-contact cuff electrode for stimulation and recording from peripheral nerves. Nerve cuffs with 16 electrodes, comprising 4 rings of 4 electrodes, were implanted around the sciatic nerve of two rats. The electromyogram signal (EMG) was recorded in response to electrical stimulation delivered by the electrodes, and the electroneurogram signal (ENG) was recorded in response to sensory stimulation applied to the ipsilateral foot. Visually detectable muscle movements were elicited with charge injections ranging from 4.6 to 8.2 nC. ENG recordings in response to sensory stimulus allowed for the onset and culmination of sensory stimulation to be detected using mean absolute value of the signal. Initial results indicate that flexion and extension of the ankle joint can be differentiated by combining information recorded from pairs of electrodes. The results of this study indicate that multi-contact cuffs can be used for decoding neural signals; however, more data needs to be collected for classification of sensory movements to be tested.**

***Index Terms*—Neural stimulation, nerve cuff, peripheral nervous system, sciatic nerve.**

I. Introduction

Interfacing with the peripheral nervous system (PNS) has long been considered as a way to restore sensorimotor deficits in patients with spinal cord injuries, amputated limbs or brain injuries [1]. The exchange of information across the neural interface can occur in two directions via stimulation or recording from nerve fibers [2]. Direct stimulation of nerve fibers via electrodes implanted in peripheral nerves has been shown to: 1) contract muscles that are no longer controlled by the central nervous system [3]; and 2) provide artificial sensory information to prosthetic hand users [4]. On the other hand the ability to record and decode sensory information from peripheral nerves will allow for closed loop control of neuroprotheses that stimulate different muscles to elicit contractions and result in functional movements of the limb [5].

One of the limitations to developing a successful PNS prosthesis is the interface between the electrodes and the nervous system [6]. Various approaches to interface with the nervous system have been developed; however, there remains a trade off between the invasiveness of the electrodes, and their ability to selectively record or stimulate nerve fibers. For stimulating PNS prostheses to be successful electrodes must be able to elicit graded sensory or motor functions, within appropriate damage limits. For recording PNS prostheses to be successful, the recorded sensory information must be able to be reliably decoded in regards to type and strength of stimulus.

Generally interfaces can be divide into two types: extraneural and intraneural electrodes [6]. Extraneural electrodes are placed around the nerve, are less invasive, and consequently less selective. Intraneural electrodes penetrate the nerve, are more selective, but also more invasive. Multi-contact nerve cuffs belong to the former group and have been developed with the hope of increasing nerve fiber selectivity while minimizing invasiveness [7]. Multi-contact cuff electrodes have been shown to allow for independent and graded control of dorsiflexion and plantarflexion [7]; and to be capable of differentiating neural signals based on the conduction velocity of action potentials [8].

In this paper we report preliminary results from the use of a commercially available multi-contact nerve cuff electrode for recording ENG and stimulation of the sciatic nerve in rats. A block diagram for our setup is shown in Fig. 1A. Importantly we examine the interactions between electrodes stimulated simultaneously and whether sensory signals could be decoded using simple features recorded from pairs of electrodes.

II. Method

A. Nerve Cuff

The concentric nerve cuff was manufactured by Microprobes (Gaithersburg, MD, USA). It consisted of 16 electrode contacts (4 rings of 4 Pt electrodes) with surface areas of approximately 0.0629 mm^2 mounted on silicon rubber tubing. The inner diameter of the cuff was 1 mm. The rings were spaced 0.75 mm apart and the distance from the end of the contacts to the end of the cuff was 1 mm. In both animals 15 of the 16 electrodes were available for both stimulation and recording. The relative position of each of the electrodes within the cuff is shown in Fig. 1B.

All animal care and procedures were performed under appropriate licenses issued by the UK Home Office under the

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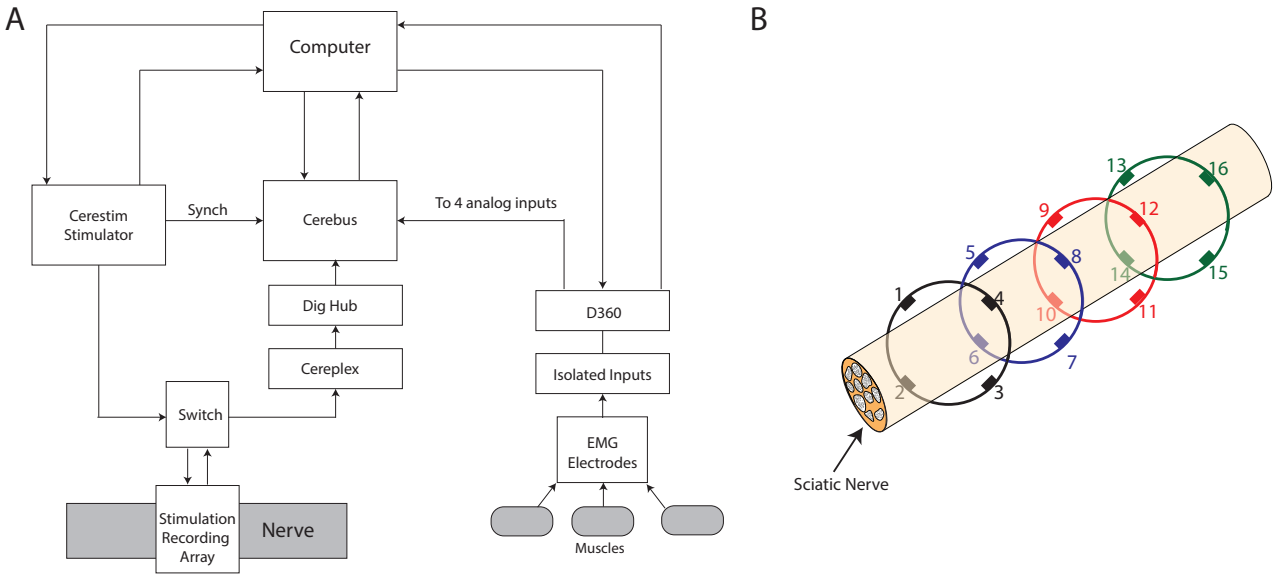

Fig. 1. A) Recording and Stimulation system setup; B) Relative positions of the electrodes in the rings within the cuff, electrodes 1, 5, 9 and 13 are on the same side of the cuff along its length and electrodes 1, 2, 3 and 4 are contained within the same ring.

Animals (Scientific Procedures) Act (1986) and were approved by the Animal Welfare and Ethical Review Board of Newcastle University.

Two Sprague Dawley rats weighing 450 and 470 g respectively were used for this study. Following the procedure introduced in [9], induction of anesthesia was performed via an intraperitoneal injection of a combination of midazolam and hypnorm at 0.27 ml per 100 g. Anesthesia was maintained by isoflurane in oxygen delivered through a nose cone as needed (between 0.5 and 1.5%) and intraperitoneal injections of midazolam and hypnorm. Fluids were maintained through a tail vein cannula given at 0.2 mL per hour.

Under anesthetic a 3 cm incision was made in the skin beginning 1 cm distal from where the femur attaches to the hip and finishing at the knee. The muscle was then carefully dissected to expose the sciatic nerve. Finally, the sciatic nerve was freed from the surrounding tissue.

A second incision was made in the skin caudal to the first on the ipsilateral side and a tunnel was made under the skin between the two incision sites. The biceps femoris was dissected away from the gluteus maximus muscle and the nerve cuff was then tunneled under the skin and the muscle to the nerve. This was done to reduce the propensity for the nerve cuff to move resulting in compression or stretch to the nerve. The cuff was then opened and placed around the nerve proximal to the point where the nerve branches into the tibial, peroneal and sural nerves and sutured into place. Kwik-Cast (World Precision Instruments, FL, USA) was placed over the cuff and the nerve to further secure the cuff in place. The muscles and skin were then sutured close to prevent the nerve and surrounding muscle from drying out.

A 1.5 cm incision was made in line with the top of the hip above L6 spinous process along the length of the spinal cord. The tissue surrounding the lamina was then carefully removed exposing the bony surface. A tungsten ground wire was then wrapped around the spinous process and secured in place with dental acrylic. For ENG recordings a second reference wire was sutured in the skin above the site of the nerve cuff implantation.

In both animals EMG was recorded from the tibialis anterior, the lateral head of the gastrocnemius, and the long digital extensor muscles. In Animal 1 recordings were also made from the medial head of the gastrocnemius and in Animal 2 from the plantaris. For bipolar EMG two tungsten wires were implanted approximately 2 mm apart into each muscle.

Following the termination of the experiments, animals were humanely killed without recovering from the anaesthesia.

B. Stimulation and Recording

A CereStim R96 (Blackrock Microsystems, Hanover, Germany) was used to deliver single symmetric biphasic current pulses with pulse widths of 200 μs and inter stimulus interval of 100 μs via a custom Matlab script. To find the threshold the stimulation current was initially stepped up and down at intervals of 5 μA, beginning at 40 μA. Once close to threshold the step size was decreased to 1 μA. Threshold was determined as the current where a visually detectable muscle twitch was elicited in two out of three trials. Once threshold was found 10 pulses were delivered to the electrode with a frequency of 1 Hz; at 80, 100, 120 and 140% of threshold. Pairs of electrodes were also stimulated with 10 pulses at 1 Hz. Current was delivered simultaneously to the two electrodes at 25, 50, 80, 100, 120 and 140% of their respective individual thresholds, i.e., if the threshold of E_1 was 30 μA and the threshold of E_2 was 40 μA at 80% of threshold E_1 would be stimulated with 24 μA and E_2 would be simultaneously stimulated with 32 μA.

EMG was measured using tungsten wire (Advent Research Materials, UK). Myoelectric signals were amplified (D360 Amplifier, Digitimer, Hertfordshire, UK) with gain of 100,

bandpassed filtered between 30 Hz and 1 kHz and subsequently sent to an analog input of a Cerebus Neural Signal Processor (Blackrock Microsystems, Hanover, Germany). The built in synch signal from the Cerestim R96 was connected to an analog input channel of the Cerebus Neural Signal Processor.

The ENG signals were recorded only in Animal 1 using the same Cerebus Neural Signal processor and measured from 15 of the 16 electrode contacts on the nerve cuff. The ENG signals were recorded at 30k samples per second, analog filtered between 0.3 and 7.5 kHz and subsequently digitally filtered between 250 Hz and 5 kHz. ENG signals were recorded in response to five different types of sensory stimulation applied to the ipsilateral foot: ankle flexion, ankle extension, outer toe pinch, thumb pinch, and stroking the dorsum of the foot. Each stimulus was repeated at least five times. The beginning and culmination of sensory stimulation were marked with a comment using the Central Suite software.

III. Results

A. Stimulation of Single Electrodes

The threshold to evoke a visually detectable muscle movement was found for 15 out of the 16 electrodes on the nerve cuff as electrode 1 was not connected. The same nerve cuff was used in both animals. The currents required to evoke a muscle movement ranged from 29 to 41 μA in the first animal and from 23 to 38 μA in the second (Fig. 2A). At threshold muscle twitches or EMG deflections were seen in only 1 or 2 muscles at most. Threshold-level stimulation on occasions resulted in the movement of a single digit.

EMG signals were recorded in response to sciatic nerve stimulation and aligned with the synch signal from the stimulator and averaged. Figure 2B shows an example of the averaged evoked EMG from the gastrocnemius muscle in response to stimulation of electrode number 2 at 80, 100, 120 and 140 % of threshold. The area under the averaged rectified EMG curve was then determined using the Trapz function in Matlab. Figure 2C shows the area under the rectified EMG curve from the gastrocnemius muscle versus the percentage of the threshold of stimulation on a single electrode.

B. Stimulation with Pairs of Electrodes

Pairs of electrodes were stimulated simultaneously with 25, 50, 80, 100, 120 and 140 % of the respective thresholds of the individual electrodes. As direct summing of the rectified EMG would overestimate the background EMG response [10], the area under the EMG curve was measured for the same time frame (20 ms) prior to the onset of the stimulus, and this was taken away from the area under the EMG curve following the stimulation. Figure 3 shows the sum of the response of the individual electrodes versus the two electrodes simultaneously stimulated at threshold. It can be seen that simultaneous stimulation of the electrodes resulted in larger EMG response than that expected from the linear sum of the electrodes individually. At 100% of threshold the response has reached a maximum, and a further increase in stimulus amplitude does not result in a larger muscle contraction. This was the case for all electrode pairs.

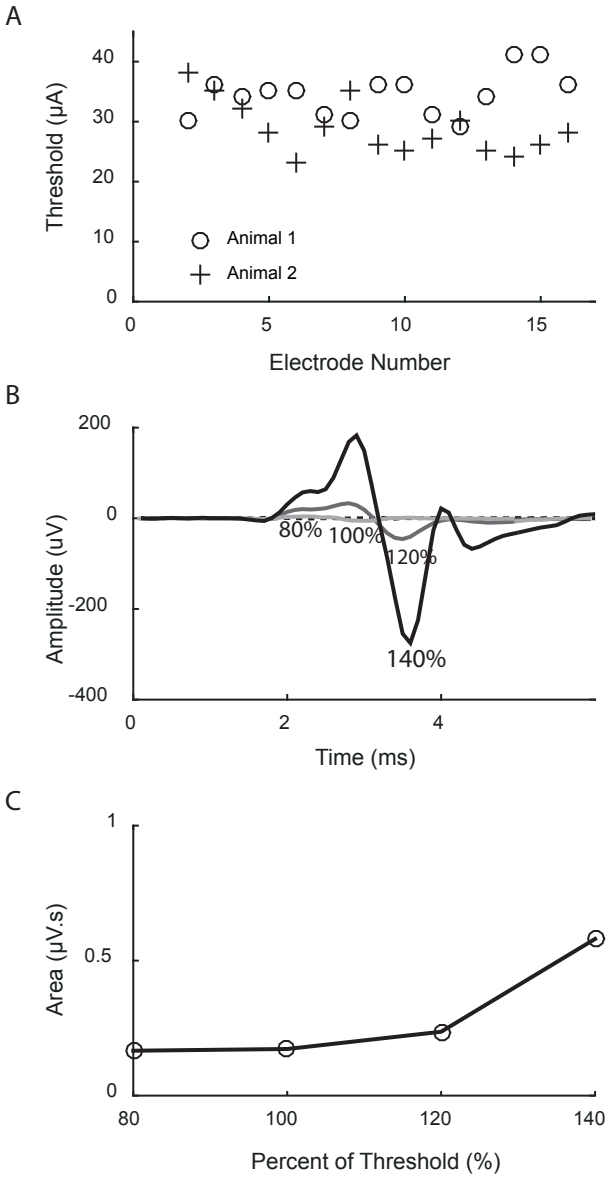

Fig. 2. A) Thresholds to elicit a visually detectable muscle twitch in Animal 1 (circles) and Animal 2 (plus signs); B) Averaged evoke EMG response to stimulation currents of 80 (dashed line), 100 (light grey), 120 (dark grey) and 140% (black) of threshold; C) Area under the curve of EMG responses shown in B.

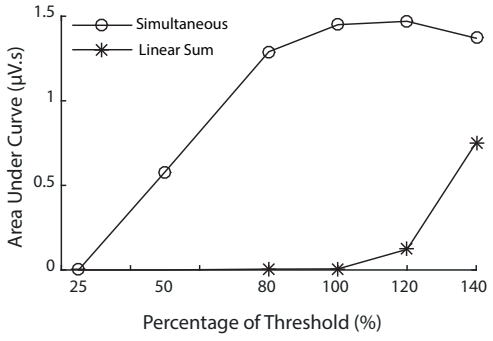

Fig. 3. A comparison of simultaneous stimulation of two electrodes with the linear sum of their individual responses. Shown for stimulation current levels of 25, 50, 80, 100, 120 and 140% of their respective individual thresholds.

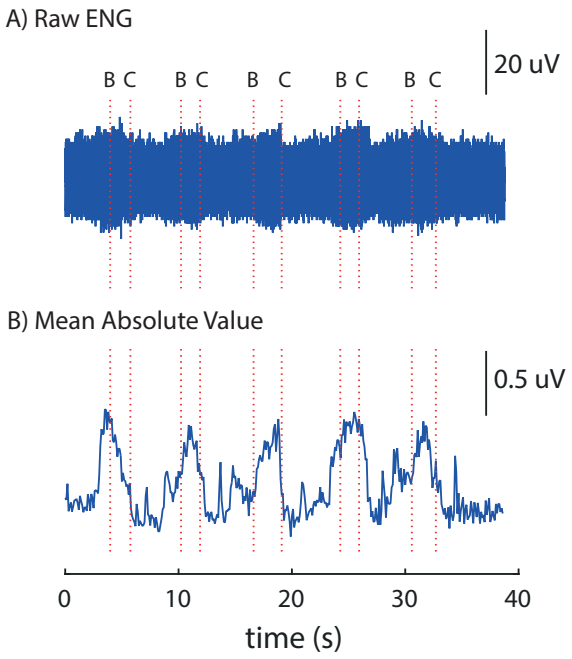

Fig. 4. A) The raw ENG signal recorded in response to flexion of the foot; B) The MAV of the signal shown in A with moving window of 100 ms. The dashed lines indicate the beginning (B) and cessation (C) of the applied sensory stimulus.

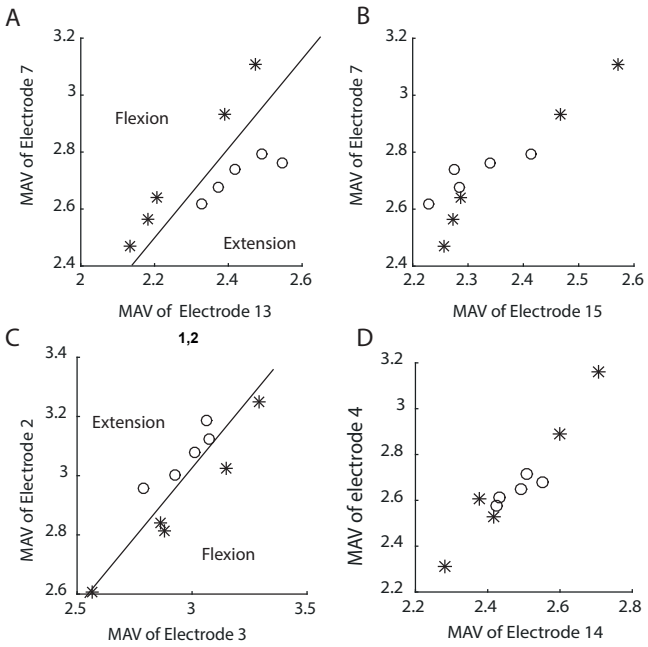

Fig. 5. Scatter plot of the MAV on pairs of electrodes from five trials of each flexion (stars) and extension (circles) of the ankle joint. (A) Electrode 7 versus electrode 13, (B) electrode 7 versus electrode 15, (C) electrode 2 versus electrode 3, and (D) electrode 4 versus electrode 14.

C. Recording of Sensory Information

The ENG was recorded in response to sensory stimulation of the foot. Fig 4A shows an example of the raw signal recorded on a single electrode in response to flexion of the ankle. The mean absolute value was calculated with a moving window of 100 ms as used in [5] (Fig. 4B). The start and cessation of sensory stimulation can be clearly identified using the mean absolute value (MAV) of the signal. Fig. 5 shows scatter plots of the MAV recorded in response to flexion and extension of the ankle from different pairs of electrodes electrodes. It can be seen that using the electrode pairs 7 and 13; and 2 and 3, the different stimuli can be easily classified along the diagonal(Fig. 5A and C). By comparison using electrode pairs 7 and 15; and 4 and 14 classification of flexion and extension could not be performed based on MAV alone (Fig. 5B and D). In this initial analysis only flexion and extension of the ankle could be clearly separated along the diagonal on the scatter plots. Of the 105 pairs of electrodes that could be chosen, 25 demonstrated clear separation between flexion and extension along the diagonal.

IV. Discussion

A. Safety of Stimulation

The maximum threshold required to evoke a muscle twitch was 41 μA in Animal 1 on both electrodes 14 and 15. This corresponds to a charge density of 13 μC/cm^2 per phase, which is well below the safe limit for neural stimulation of 224 μC/cm^2 calculated using the relationship developed by R.V. Shannon of,

$$\log\left(\frac{Q}{A}\right) = k - \log(Q) \quad (1)$$

where A is the surface area in cm^2, and Q is the charge in μC [11]. This demonstrates that these electrodes are capable of delivering effective stimulation well within established safe limits.

B. Stimulation with Pairs of Electrodes

Simultaneously stimulation of any two electrodes on the cuff results in a supralinear response regardless of their relative positions on the cuff. Supralinear responses with paired electrode stimulation have previously been seen in the sciatic nerve of cats using Utah arrays [12] and are not unexpected. If the electric fields of the two electrodes used for stimulation overlap, fibers that lie in an area where the electric field was not large enough for them to fire in response to a single electrode will now receive extra charge from the second electrode and fire. If these cuffs are to be used in motor or sensory stimulating neuroprostheses, further investigation of interactions between the electrodes is needed. Importantly it should be determined how far apart pulses would need to be spaced to avoid electrode interactions, and how electrode interactions may affect either the direction of movement of the leg/foot in motor prostheses or the sensation felt in sensory prostheses.

C. Sensory Recordings

Previously Raspovic *et al*. demonstrated that sensory information obtained from mechanical stimulation of the foot could be decoded from ENG signals recorded with single channel cuff electrodes [5]. Additionally Zariffa *et al*. has demonstrated that discrimination of signals originating via electrical stimulation of one of three branches of the sciatic

nerve (tibial, peroneal, or sural branches) can be improved by using a matrix of electrode contacts placed around the nerve [13]. Furthermore, they showed that this improvement in discrimination was due to being able to choose the most informative electrodes from a large number, and that the most informative electrode location could not necessarily be predicted prior to electrode implantation and experimentation.

Building on these studies, here we have examined discrimination of different types of mechanical stimulation of the foot using a matrix of 16 electrodes placed around the sciatic nerve on a cuff. Using the simple measure of MAV on the different electrodes we have shown that flexion and extension of the ankle could be clearly separated using information obtained from as little as two electrodes (Fig. 5). In agreement with [13] the most informative electrodes could not be predicted prior to performing the experiment. This is evident as in some cases electrodes on the same ring, such as electrode 2 and 3 (Fig. 5C), showed clear separation on scatter plots, whereas electrodes 4 and 14 (Fig. 5D) that were located far apart on opposite sides of the cuff did not. Nevertheless, we have demonstrated the potential for different sensory signals to be discriminated using extraction of simple features recorded from pairs of electrodes. Further analysis is needed to determine if other sensory modalities can be discriminated.

D. Limitations

The scatter plot shown in Fig. 5 shows that flexion and extension of the leg could be classified using MAV recorded on a single pair of electrodes. ENG signals were only recorded in one animal for 5 trials of each signal type. For classification algorithms to be tested more trials of each sensory stimulus will be needed. Additionally, these recordings were taken in anaesthetized animals, and we did not have to worry about interference of other biological signals such as EMG. Future research is required to determine whether sensory information can still be obtained in awake and moving animals.

V. Conclusion

The results of this preliminary study indicate that the multicontact nerve cuff used would be capable of delivering effective stimulation to elicit muscle contractions within safe stimulation levels. Additionally, it shows that the simple feature of MAV recorded from only two electrodes of a multichannel nerve cuff can be used to decode sensory information. Future work may include decoding of different sensory stimuli via statistical analysis of multichannel ENG.

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